SYSTEM FOR PHASE INVERSION ULTRASONIC IMAGING

CROSS-REFERENCE TO RELATED APPLICATIONS

[0001]   This application is a continuation—and claims the priority benefit—

of commonly owned United States patent application number 09/872,541,

entitled "Method for Phase Inversion Ultrasonic Imaging," filed May 31, 2001.

The subject matter of this application is incorporated herein by reference.

[0002]   This application is related to commonly owned United States patent

application number 10/226,843, entitled "Method for Phase Inversion Ultrasonic

Imaging," filed August 21, 2002. U.S patent application number 10/226,843 is a

divisional of U.S. patent application 09/872,541. The subject matter of this

application is incorporated herein by reference.

BACKGROUND OF THE INVENTION

Field of the Invention

[0003]   The present invention relates to ultrasound imaging, and more

particularly, to a system for improving and enhancing ultrasound images.

Description of the Background Art

[0004]   Ultrasonic imaging is frequently used for a variety of diagnostic

procedures due to its non-invasive nature, low cost, and fast response time.

These qualities are especially true in medical fields where the added benefit is

reducing or eliminating a patient's exposure to radiation. Typically, ultrasound

imaging is accomplished by 1) generating and directing an ultrasonic beam into

media under investigation; and 2) observing any resulting waves that are

reflected back from dissimilar tissues and tissue boundaries within that area.

The resulting waves are received as signals. These received signals are then post-

processed and imaged on a screen by plotting a spot whose intensity is

proportional to the amplitude of a reflected beam from a given location.

Determination of location is based upon a known transmission and re-radiation

rate after the ultrasonic wave is pulsed into the media under investigation.

[0005]   Typically, an ultrasonic signal transmitted into the media under

investigation includes a burst of sinusoidal waves of a given waveform. These

sinusoidal waves are applied to a transducer and form a transmitted signal. The

transmitted signal is typically in the range of 40 kHz to 50 MHz, but more

commonly, in the range of 40 kHz to 1 MHz. As the transmitted signal interacts

with tissue layers and boundaries between layers, the ultrasonic signal is

modified by being scattered, resonated, attenuated, reflected, or transmitted.

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[0006]   Media under investigation are often a non-linear media such as those commonly found in the human body. Non-linear media produce harmonic frequencies in the echoed signals. These additional frequency components continue to re-radiate through and, in turn, reflect off or interact with other structures. A portion of the reflected (or echoed) signals propagates back to a receiving transducer.

[0007]   Fundamental and harmonic frequencies that are impinged upon a receiving transducer includes the full signal, which must then be further processed to eliminate noise and extraneous components. The receiving transducer may be the same as a transmitting transducer, or can be completely independent. When the same transducer is used, a transmit/receive (T/R) switch connects the transducer to either the transmitter electronics or the receiver post-processing electronics. The receiving transducer accepts the echo signal plus any generated noise and furnishes these to a portion of the post-processing electronics known as a beam former. Beam formers reject noise and have either an adaptive or fixed configuration. Adaptive beam formers are designed to reject variable directional noise sources by monitoring the noise field and adjusting internal parameters to minimize the background noise. Fixed beam formers are designed to reject isotropic noise and take advantage of the directional property of the reflected signal.

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[0008]   Ultimately, ultrasonic images of the human body are a product of harmonic imaging. Harmonic imaging is generally associated with either visualization of tissue boundaries and densities of different media, or imaging contrast agents at harmonic frequencies. Contrast agents are typically fluid filled micro-spheres that resonate at ultrasonic frequencies. Such agents are injected into the blood stream and are carried to various parts of the body. Once these agents are pulsed at ultrasonic frequencies, harmonic echo-locator signals are generated due to the resonance produced within the micro-spheres.

[0009]   While ultrasonic procedures have a distinct number of advantages over other types of diagnostic techniques, prior art methods and systems have noise problems that make it difficult to determine the exact location and proper interpretation of the received signal. Various forms of averaging techniques have been employed to reduce the noise, but averaging alone is ineffective in locating images of interest between tissues with similar densities. (Echoed signals from tissues with similar densities will indicate a uniform mass with indistinct boundaries. Averaging won't help in this situation.) These interpretation difficulties are exacerbated by the fact that many tissues in the human body have similar densities. Therefore, a method and system are needed that can effectively overcome the stated difficulties while not negating the positive benefits of ultrasound imaging systems in general.

SUMMARY OF THE INVENTION

[0010]   The present invention is a system and method for generating enhanced ultrasonic images. The invention utilizes multiple ultrasonic pulses that are transmitted in an alternating fashion into media of interest. These media being imaged may be a human body or some other linear and/or non-linear media. The ultrasonic pulses are modulated in a way that may vary in amplitude, frequency, phase, or pulse width. Each set of ultrasonic pulses is out-of-phase with other ultrasonic pulses by $\dfrac{360°}{k}$, where k is the number of pulse sets in the pulse sequence for a given transducer element number, n. An out-of-phase condition is a property when waveforms are of the same frequency but do not have corresponding intensity values at the same instant. The echo signals generated by the non-linear media interacting with these out-of-phase signals are measured and appropriately combined.

[0011]   The present invention is based on the observation that many types of media scatter sound in a non-linear manner. With an ultrasonic imaging system based on linear-scattering media, the return signal is a time-shifted, amplitude-scaled version of the incident signal. Non-linear scattering media produce signals that cannot be produced by simple time-shifts, scaling, or summation of the signal incident to a scattering site. The phase of an ultrasound wave reflected

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from the boundary of a non-linear medium is altered in a manner that depends on the phase of the incident sound pulse. For example, consider the special case of two ultrasound pulses ($k = 2$), where the phase difference between the two transmitted sound pulses differ by $\dfrac{360°}{k} = \dfrac{360°}{2}$ or 180 degrees. If the scattering site were purely linear then the received signal from each of the transmitted pulses would be the inverse of each other. These inverse signals, when averaged, would have a sum of zero. If, however, there are signals generated from a non-linear process within the media, then these signals will not be the inverse of each other and, consequently, will not sum to zero.

[0012]   This non-linear property can be used to construct a system that looks at the non-linear regions within given media. For example, in one embodiment of this system, an image area is formed from three different sets of transmitted signals each differing in phase by 120 degrees. The linear reflections generated by these $k = 3$ sets of excitation pulses will cancel each other, while the kth order non-linear components will not. This pulse cancellation allows an averaged set of raw received data, $F(n, t)$, to be produced. A three-dimensional averaged set of raw received data, $F(n,m,t)$, may also be produced where m is an element from a two-dimensional, or $n \times m$ transducer array. The data functionality is dependent on the channel (or transducer) number and time. This single set of

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averaged data can then be used to re-construct an image area. The image area

reconstructed would represent the information generated from the third, or k^{th},

harmonic, generated from the scattering sites within the media.

BRIEF DESCRIPTION OF DRAWINGS

[0013]   FIG. 1 is a schematic block diagram of an embodiment of an

ultrasound imaging system using the present invention;

[0014]   FIG. 2 shows an exemplary modification of one of a potential

plurality of waveforms being modified;

[0015]   FIG. 3 shows an embodiment of a signal transmitter unit prior to

signal delivery to media of interest;

[0016]   FIG. 4A shows an embodiment of a receiver and raw data averager

unit;

[0017]   FIG. 4B shows an alternative embodiment of a receiver and raw data

averager unit; and

[0018]   FIG. 5 shows an embodiment of a data processing unit.

DETAILED DESCRIPTION OF THE INVENTION

[0019]   The present invention relates to the field of ultrasound imaging. This invention utilizes broad beam technology (B² Technology™) to perform image extraction of the non-linear elements of media under investigation. These media will hereinafter be referred to as media of interest. Broad beam technology defines an area under investigation at a given point in time, which is in contrast with a system utilizing a focused beam.

[0020]   FIG. 1 is a block diagram of an embodiment of an ultrasound imaging system using the present invention. Imaging system **100** includes at least one signal generator unit **110**, at least one signal transmitter unit **120**, media of interest **130** to be imaged, at least one receiver and raw data averager unit **140** to capture signals received from the media of interest **130**, and a data processing unit **150** for taking the averaged received signals and producing an area of image formation on an image display unit **160**.

[0021]   A signal generator unit **110** drives circuitry for a signal transmitter unit **120**. The signal transmitter unit is shown and described in greater detail in FIG. 3.

[0022]   A signal transmitter unit **120** transmits pulsed sets of ultrasonic energy into the media of interest **130**. Echoes received from the media of interest

130 are stored in the receiver and raw data averager unit **140**. Subsequent out-of-phase signals from the signal generator unit **110** pass through the signal transmitter unit **120**, and are converted into pulsed sets of ultrasonic energy that travel to the media of interest **130**. The media of interest **130** modifies the pulsed sets of ultrasonic energy. These modified pulsed sets of ultrasonic energy are received and averaged by the receiver and raw data averager unit **140**. Data from these received pulsed sets are averaged in a data set as a function of channel number, n, and time, t. The averaged data sets are processed by the data processing unit **150** and displayed on an image display unit **160**.

[0023]   FIG. 2 shows an exemplary modification of one of a potential plurality of waveforms being modified. This FIG. 2 example demonstrates how a signal generator unit **110** may modify a generated signal. For example, a stated modulation may be in the form of any combination of varying an amplitude, frequency, or pulse width of an unaffected input signal. These modulated signals will additionally vary in phase for a given pulse set. To produce a modulation, an envelope function, $A(n,t)$, may be convolved with a sine wave, depicted by $e^{j[\omega_o(t,n)t+\theta_i+\theta(n)]}$ giving the final waveform $A(n,t)\ e^{j[\omega_o(t,n)t+\theta_i+\theta(n)]}$. In this waveform notation, n is the transducer element number, and i is a given pulse index (e.g., if a second harmonic is utilized, $k = 2$, then $i = 1 .. 2$). The phase

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varies for different pulse sequences within a given pulse set and is indicated by the θ_i notation.

[0024]   To illustrate the concept of phase variation, take an example where k is three. In this example, each pulse within a pulse set is varied in phase by $\frac{360°}{k} \equiv \frac{360°}{3}$ or 120°. A first pulse is generated with a 0° phase orientation, a second pulse is 120° out-of-phase with the first pulse, and a last pulse in the pulse set is 240° out-of-phase with respect to the first pulse. After the first pulse is transmitted and received, the second pulse is transmitted and received, and so on through the sequence. All information is tracked so that fundamental frequencies can be summed and eliminated, leaving primarily only harmonically generated echoes. Recall that harmonically generated echoes are produced by non-linear media.

[0025]   Further, as an example, an envelope function, $A(n,t)$, may be a Gaussian waveform. The transmitted signal may additionally be modulated as a chirped waveform (i.e., swept-frequency modulation, a Fourier transform of which is still centered around the fundamental with a broader dispersion). Optionally, a digital waveform generator could be used in place of the convolution method shown in FIG. 2.

[0026]   FIG. 3 shows an embodiment of a signal transmitter unit **120** prior to

signal delivery to media of interest. The signal transmitter unit **120** includes at

least a power amplifier **330**, a transmit/receive switch **340**, and a first transducer

350. Optionally, a signal transmitter unit **120** may further include a delay circuit

310. The delay circuit **310** may be an analog or digital delay. Also, optionally,

the signal transmitter unit **120** may include a channel gain unit **320** to drive the

power amplifier **330** as a function of channel number and time. Additionally, the

signal or pulse may be pulse-width modulated (not shown) to conserve power.

Power conservation can become crucial in field applications of the system where

battery power may be utilized.

[0027]   FIG. 4A shows one embodiment of a receiver and raw data averager

unit **140**. A second transducer **410** receives pulsed sets modified by media of

interest **130**. These received pulsed sets are transformed from ultrasonic energy

into an electrical signal by the second transducer **410**. A second transmit/receive

switch **420** may be used to couple the electrical signals to the appropriate

circuitry. In one embodiment, a second transducer **410** and a second

transmit/receive switch **420** may be coincident or analogous units to the first

transducer **350** and first transmit/receive switch **340** shown in FIG. 3. A second

power amplifier **430** may be added and controlled as a function of time by gain

control unit **440**. The output of power amplifier **430** sends an amplified signal to

an optional bandpass filter **450**. The bandpass filter **450** may be used, among

other things, to reduce or eliminate extraneous noise. FIGs. 4A and 4B share the

same component layout up to and including bandpass filter **450**, where thereafter

they diverge thereby exemplifying alternate embodiments.

[0028]   The electrical signal of the FIG. 4A embodiment is coupled to a first

analog-to-digital (A/D) converter **460**, and may continue into an optional in-

phase and quadrature (I/Q) mixer **470**, which produces a single side-band signal,

optional first baseband filter **480**, and to an averager **490**. The optional first

baseband filter acts to reduce or eliminate any fundamental frequency from the

signals received from the original pulse sets, leaving primarily harmonically

generated signals. One purpose of the averager is to provide a point-by-point

arithmetic average of the received electrical signals. Mathematically, this

arithmetic average may be expressed as $F(n,t) = \sum_{i=1}^{k} \dfrac{R_i(n,t)}{k}$, where the received

signal, R_i, is summed for each element of the transmit cycle as a function of

channel number and time to correlate with the original transmitted pulse, i. All

other components in the signal path are of types commonly known to one of

ordinary skill in the art.

[0029]   FIG. 4B shows an alternative embodiment of a receiver and raw data

averager unit. Recall FIGs. 4A and 4B share the same component layout through

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and including the optional bandpass filter **450**. From the point of this optional

bandpass filter **450**, the signal of the FIG. 4B embodiment is further coupled to an

analog mixer **455**, an optional second baseband filter **465**, a second analog-to-

digital converter **475**, and an averager **490**.

[0030]   FIG. 5 shows an embodiment of the data processing unit **150**. Here,

data processing unit **150** receives averaged data from the receiver and raw data

averager unit **140**. The averaged data are input to the data processing unit **150**

and received at I/Q raw data matrix **510**, which stores the averaged data in an

MxN area array, where M is the number of samples (1 to 10,000 samples is an

exemplary number) and N is the number of elements x 2 (both in-phase and

quadrature). These averaged data are fed into a digital signal processor (DSP)

520, which reconstructs the raw data into an area of acoustic image. An

exemplary reconstruction equation may take the form of

$I(r,\varphi) = \sum_{i=1}^{k} a_i(r,\varphi) \cdot e^{j\theta_i(r,\varphi)} \cdot F[i,t_i(r,\varphi)]$. In this equation a_i indicates an aperture

function, r refers to a radial distance from a transducer center at a given angle φ,

and the function F is an averaged set of raw received data. The digital signal

processor **520** functions could be achieved in any number of ways, For example,

in an alternative embodiment, a properly designed application-specific

integrated circuit (ASIC) could be used in place of the digital signal processor

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520. These converted data in polar coordinates are saved in an acoustic image

data buffer 530 in a JxK matrix (where J is the number of range samples and K is

the number of angular samples). At this point, the data are still a function of a

distance, r, from the transducer at a given angle, φ. This could also be

accomplished in a Cartesian coordinate system. The acoustic image data buffer

530 allows the data to be stored until needed by scan converter 540. The

$I(r,\varphi)$ image data are converted into a reconstructed image in Cartesian

coordinate data $I(x,y)$ through the use of an r-φ scan converter 540. An r-φ scan

converter is well known in the art and typically converts two-dimensional data

from polar to Cartesian coordinates by means of the conversion,

$x = r\cos(\varphi)$ and $y = r\sin(\varphi)$.

[0031]   Output from the data processing unit 150 produces an image area

$I(x,y)$ corresponding to an area irradiated by pulsed sets of ultrasonic energy.

These converted $I(x,y)$ data may be displayed on image display unit 160. Image

display unit 160 may be any visual display such as, but not limited to, a

computer monitor, flat-panel or liquid-crystal display, cathode-ray tube (CRT),

or the like.

[0032]   From the description of the preferred embodiments of the process

and apparatus set forth supra, it will be apparent to one of ordinary skill in the

art that variations and additions to the embodiments can be made without departing from the principles of the present invention. For example, it could be easy to envision a system whereby an entire three-dimensional (3D) volume could be displayed at once as opposed to a two-dimensional area. This three—dimensional embodiment may be accomplished by holography or some other means. It would be an obvious extrapolation from the tenets of the two-dimensional system presented herein to construct a three-dimensional apparatus.